A Bidirectional Analog VLSI Cochlear Model

Lloyd Watts
Richard F. Lyon [1]
Carver Mead
Department of Computer Science
California Institute of Technology
Pasadena, CA USA 91125

Abstract

A novel circuit is presented for implementing a bidirectional passive cochlear model in analog VLSI. The circuit includes a subcircuit for modelling the fluid in the cochlear duct, and a subcircuit for modelling the passive basilar membrane. The circuit is compared to the classical 1-D transmission line cochlear model and found to be equivalent. The approach leads to an unexpected fault tolerance in the form of insensitivity to transconductance amplifier offset voltages. A 545-stage cochlea has been fabricated and demonstrates the expected wave propagation behaviour.

1 Introduction

The cochlea is a complex 3-dimensional fluid-dynamical system, which uses sensitive hair-cell transducers to detect travelling waves in a fluid/membrane structure. Over the last 60 years, models of the cochlea have been developed to explore hypotheses about this cochlear travelling wave mechanism.

Early models, such as the passive linear 1-D transmission line model [16] were successful in qualitatively explaining the travelling wave behaviour observed in cadaver ears [12]. However, when low-amplitude data from living animals became available in the early 1970's [10], it became clear that the passive linear 1-D model was inadequate. During the 1970's and early 1980's, research focussed on passive linear models of higher dimensionality [1, 11]. Again, it was found that passive linear models of 2 or 3 dimensions were unable to explain quantitatively both the amplitude and phase characteristics of the travelling wave mechanism.

Evidence of the need for an active model had been accumulating with the observation of tinnitus (sustained ringing in the ears) and otoacoustic emissions (sounds emanating from the ear) [6], while the need for a nonlinear model was supported by two-tone suppression experiments and the observation of distortion product tones [14]. The hearing research community

[1] R. F. Lyon is also with Apple Computer Inc., Cupertino, CA 95014.

now generally accepts the Outer Hair Cells (OHCs) as the active nonlinear process in the biological cochlea, and the search has been under way since the early 1980's for a comprehensive and quantitative explanation of their role in the hearing process [2].

In the simpler models, such as the passive linear models, it is possible to obtain approximate closed-form solutions for the travelling wave [13, 4]. For the more complex active nonlinear models, closed-form solutions will be very difficult to find except under very severe restrictions of form. For this reason, most researchers have turned to digital computer simulation as their primary tool in investigating their active nonlinear models.

The disadvantage of digital computer simulation of a cochlea is that a very fine spatio-temporal quantization is necessary to ensure convergence to a solution, resulting in very long simulation execution times. For this reason, a real-time analog VLSI model was proposed [7], using a unidirectional cascade of second-order filter sections (the "unidirectional filter cascade" or UFC model). The UFC model has the advantage of using real-time analog components with no time quantization (like a real cochlea), and can exploit the massive integration scale available from today's semiconductor manufacturers to include many processing stages on a single chip, resulting in a very fine spatial quantization. The UFC model qualitatively replicates the observed behaviour of real cochleas, namely that waves normally propagate without reflection [3] with exponentially decreasing velocity and wavelength [8], until they sharply cut off. An active mechanism was included in the UFC model to model the active Outer Hair Cell behaviour. The unidirectionality of the UFC model provided an effective engineering solution to stability problems in the presence of these active components. Unfortunately, this unidirectionality makes it impossible to use the UFC model to investigate one of the most pressing questions of cochlear modelling today: How does the active *bidirectional* travelling wave structure of the biological cochlea remain stable?

As a first step towards addressing this question, we present a passive bidirectional cochlear model suitable for implementation in analog VLSI.

2 The Cochlear Model

The wave equation which governs the 1-D passive cochlea is [5]:

$$\frac{d^2P(x)}{dx^2} = -\frac{2\rho\omega^2}{h(S(x) + i\beta(x)\omega - M(x)\omega^2)}P(x) \tag{1}$$

where P is the pressure in the cochlear fluid, x is the position along the cochlea, ρ is the density of the cochlear fluid, ω is the frequency of the stimulus, h is the height of the duct, $S(x)$ is the stiffness of the membrane at the given position, $\beta(x)$ is the damping of the membrane at the given position, and $M(x)$ is the mass of the membrane at the given position. The

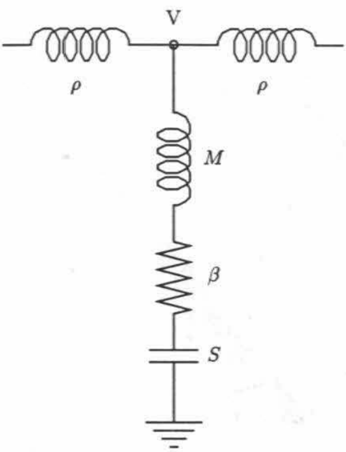

Figure 1: One section of the classical 1-D transmission line cochlear model.

cochlea achieves a $\log\omega \Longleftrightarrow x$ mapping by varying the stiffness, mass and damping terms exponentially along its length.

The well-known transmission line model [16] shown in Figure 1 is governed by a wave equation of the same form as equation (1), and thus has long been a useful model for engineers familiar with transmission line analysis techniques. This type of model has actually been constructed using discrete components [15]. However, the transmission line model is inappropriate for integrated circuit implementation because of the inductors. It would be possible to replace the inductors by an equivalent circuit, however it is much more efficient to go back to the physics of the cochlea and derive a new circuit which is appropriate for the analog VLSI medium.

Such a circuit is shown in Figure 2. The wave equation for this circuit is

$$\frac{d^2V(x)}{dx^2} = -\frac{2R\omega^2}{S(x) + i\beta(x)\omega - M(x)\omega^2}V(x) \tag{2}$$

where

$$S(x) = \frac{g_1(x)}{C_1C_2} \tag{3}$$

$$\beta(x) = \frac{1}{C_2} \tag{4}$$

$$M(x) = \frac{1}{g_2(x)} \tag{5}$$

and the fluid mass is represented by the resistance R. The voltage $V(x)$ represents the velocity potential of the fluid (proportional to pressure for a given frequency) and the current $I(x)$ represents the velocity of the basilar membrane. Clearly, this wave equation has the same form as equation (1),

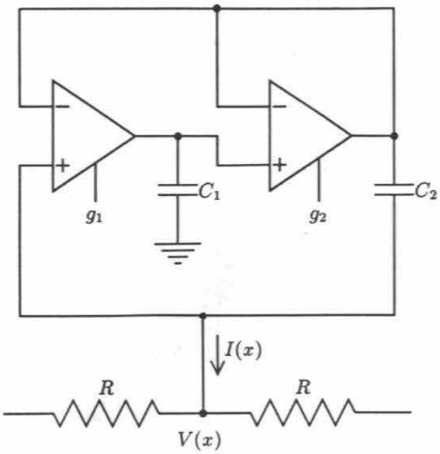

Figure 2: One section of the passive bidirectional VLSI cochlear model.

and thus the circuit is indeed a 1-D cochlear model. In practice, the resistor R may be made by using, for example, the "HRes" circuit described in [9].

As with the Unidirectional Filter Cascade model [7], the transconductance amplifiers are intended to be operated in the subthreshold regime, where there is an exponential relationship between the bias voltage and the transconductance g of the amplifier. This allows a linearly decreasing bias voltage ("tilted control line") to effect an exponentially decreasing transconductance g, which is required to achieve the $\log \omega \Longleftrightarrow x$ mapping. The linearly decreasing bias voltage as a function of distance along the cochlea is achieved by using a long polysilicon wire to bias the amplifiers, and applying a voltage difference across its two ends.

One feature deserves special note: there is some "fault tolerance" inherent in this circuit. The capacitive coupling between the fluid subcircuit and the membrane subcircuit ensures that the wave propagation is not affected by the DC operating point of the membrane subcircuit. For this reason, the model is insensitive to transconductance amplifier voltage offsets in the membrane subcircuits. In addition, the circuit is expected to be tolerant of some number of failed membrane subcircuits (stuck-on or stuck-off), since the wave can still propagate through the fluid mass portion of the circuit.

3 The Test Chip

The test chip consisted of a 545-stage cochlea, in five rows of 109 stages, with a serial-analog-multiplexer, or "scanner" to allow monitoring of the "velocity potential" voltages and the "membrane velocity" currents through two pins. The first stage of the cochlea is at the lower left corner of the chip,

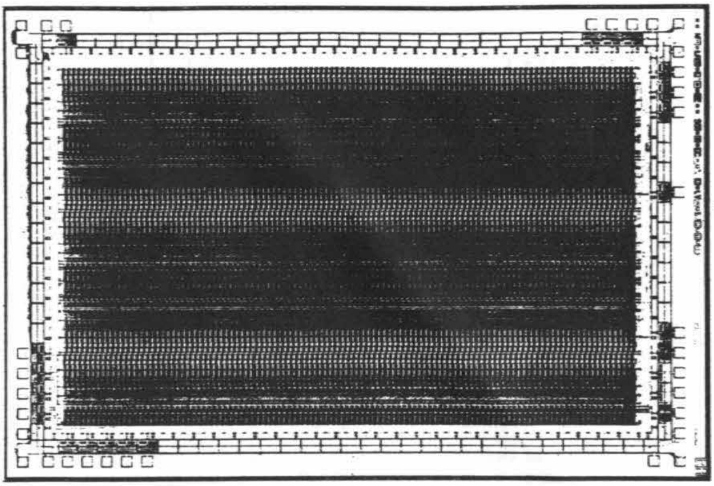

Figure 3: Photograph of the bidirectional cochlea test chip.

and the last stage is at the upper right. A photograph of the chip appears in Figure 3. The dimensions of the chip were $4600\mu m \times 6800\mu m$. The chip was fabricated on a p-well double-poly process.

4 Test Results

The test chip exhibited the expected wave propagation behaviour. By applying suitably tilted bias voltages to the cochlear model, it was possible to observe a slowing and shortening of the wave as it travelled down the cochlea. The structure is functionally symmetrical, and so of course waves propagate in both directions. With different bias settings, it was possible to propagate waves over the entire frequency range of human hearing (up to 20 kHz).

For very low frequencies (below 10 Hz), it was possible to observe the travelling waves directly using the scanner. Several traces of the the travelling pressure and velocity waves are shown in Figures 4 through 7. For all data shown, the cochlea was tuned for frequencies between 1-8 Hz. Figures 4 and 5 show the fluid pressure and membrane velocity responses to a 2 Hz stimulus. Figures 6 and 7 show the fluid pressure and membrane velocity responses to a 6 Hz stimulus. Several phases of the travelling waves are superimposed to allow the envelope of the wave to be visualized. The pressure wave decreases in amplitude as it travels down the cochlea, while the

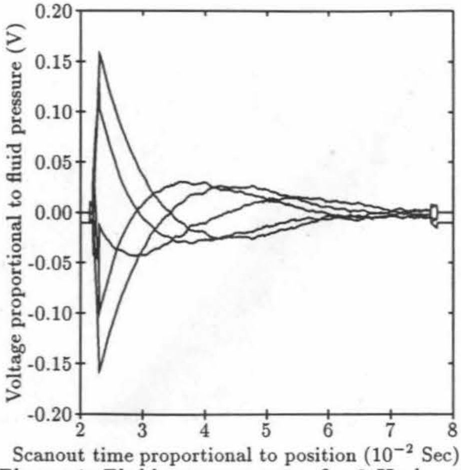

Figure 4: Fluid pressure wave for 2 Hz input.

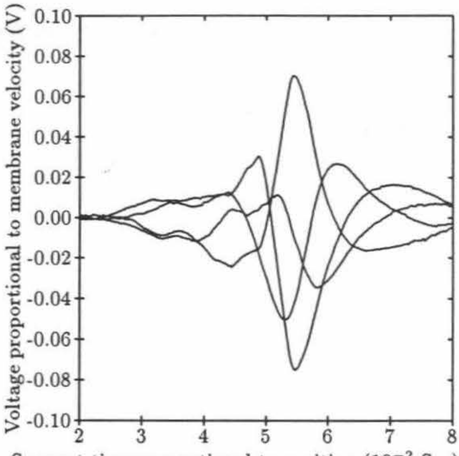

Figure 5: Membrane velocity wave for 2 Hz input.

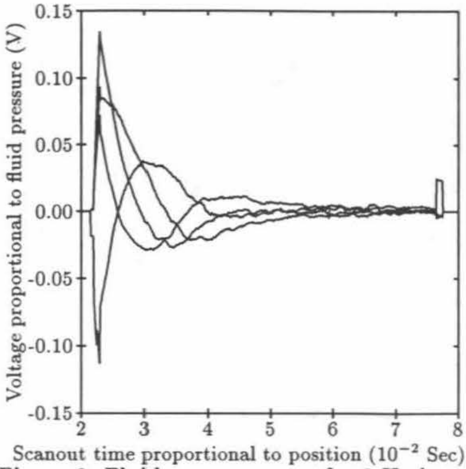

Figure 6: Fluid pressure wave for 6 Hz input.

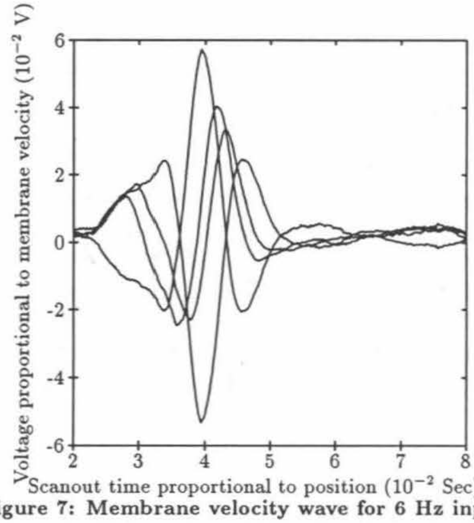

Figure 7: Membrane velocity wave for 6 Hz input.

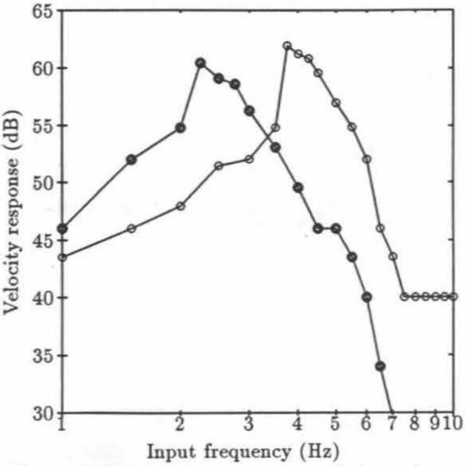

Figure 8: Frequency response at two places.

membrane velocity wave has a pronounced peak. The position of the peak depends on the stimulus frequency; low stimulus frequencies travel a long distance down the cochlea before they reach their peak and cut off, while high stimulus frequencies travel a shorter distance.

It is also possible to plot the response at a given position as a function of stimulus frequency. Such a plot is shown in Figure 8, showing the membrane velocity for two positions in the silicon cochlea. Each measurement was taken by holding the scanner fixed and sweeping the input frequency. For each position, there is a response peak at some characteristic frequency. These responses are generally as expected, although the high-frequency cutoff behaviour was expected to be somewhat steeper.

5 Conclusions

We believe that analog VLSI is an appropriate medium for investigating cochlear models, especially nonlinear ones where closed-form solutions are difficult to find, and digital simulation may be cumbersome. As a first step towards a realistic nonlinear analog VLSI cochlear model, we have implemented a bidirectional cochlear model which uses only transconductance amplifiers, resistors, and capacitors. The model is equivalent to the classical linear 1-D transmission line cochlear model. A test chip with 545 stages has been fabricated and tested, and is capable of propagating waves representing fluid pressure and membrane velocity at frequencies spanning the normal range of human hearing.

6 Acknowledgements

The authors are pleased to acknowledge many helpful discussions with Xavier Arreguit, John Lazzaro, and Massimo Sivilotti. This work was sponsored by the Office of Naval Research and the System Development Foundation. Chip fabrication was provided by the Defense Advanced Research Projects Agency and the MOSIS Service.

References

[1] J. B. Allen and M. M. Sondhi. Cochlear macromechanics: Time domain solutions. *Journal of the Acoustical Society of America*, 66:123–132, July 1979.

[2] E. de Boer. No sharpening? a challenge for cochlear mechanics. *Journal of the Acoustical Society of America*, 73:567–573, 1983.

[3] E. de Boer and R. MacKay. Reflections on reflections. *Journal of the Acoustical Society of America*, 67:882–890, 1979.

[4] E. de Boer and M. A. Viergever. Validity of the liouville-green (or wkb) method for cochlear mechanics. *Hearing Research*, 8:131–155, 1982.

[5] W. Bialek et al. *The Vertebrate Inner Ear*. Academic Press, 1982.

[6] D. T. Kemp. Stimulated acoustic emissions from within the human auditory system. *Journal of the Acoustical Society of America*, 64:1386–1391, 1978.

[7] R. F. Lyon and C. A. Mead. An analog electronic cochlea. *IEEE Trans. Acous. Speech, and Signal Proc.*, 36(7):1119–1134, July 1988.

[8] R. F. Lyon and C. A. Mead. Cochlear hydrodynamics demystified. *Caltech Computer Science Tech. Rept.*, Caltech-CS-TR-884, Feb 1988.

[9] C. A. Mead. *Analog VLSI and Neural Systems*. Addison-Wesley, 1988.

[10] W. S. Rhode. Observations of the vibration of the basilar membrane in squirrel monkeys using the mossbauer technique. *Journal of the Acoustical Society of America*, 49:1218–1231, 1971.

[11] L. A. Taber and C. R. Steele. Cochlear model including three-dimensional fluid and four modes of partition flexibility. *Journal of the Acoustical Society of America*, 70:426–436, 1981.

[12] G. von Békésy. *Experiments in Hearing*. McGraw-Hill, 1960.

[13] G. Zweig, R. Lipes, and J. R. Pierce. The cochlear compromise. *Journal of the Acoustical Society of America*, 59:975–982, 1976.

[14] E. Zwicker. Suppression and $(2f_1 - f_2)$-difference tones in a nonlinear cochlear preprocessing model with active feedback. *Journal of the Acoustical Society of America*, 80:163–176, 1986.

[15] E. Zwicker and W. Peisl. Cochlear preprocessing in analog models, in digital models and in human inner ear. *Hearing Research*, 44:209–216, 1990.

[16] J. J. Zwislocki. Theory of the acoustical action of the cochlea. *Journal of the Acoustical Society of America*, 22:778–784, 1950.